Muscle Fibers Inspired High-Performance Piezoelectric Textiles for Wearable Physiological Monitoring

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The next-generation wearable biosensors with highly biocompatible, stretchable, and robust features are expected to enable the change of the current reactive and disease-centric healthcare system to a personalized model with a focus on disease prevention and health promotion. Herein, a muscle-fiber-inspired nonwoven piezoelectric textile with tunable mechanical properties for wearable physiological monitoring is developed. To mimic the muscle fibers, polydopamine (PDA) is dispersed into the electrospun barium titanate/polyvinylidene fluoride (BTO/PVDF) nanofibers to enhance the interfacial-adhesion, mechanical strength, and piezoelectric properties. Such improvements are both experimentally observed via mechanical characterization and theoretically verified by the phase-field simulation. Taking the PDA@BTO/PVDF nanofibers as the building blocks, a nonwoven light-weight piezoelectric textile is fabricated, which hold an outstanding sensitivity (3.95 V N⁻¹) and long-term stability (<3% decline after 7,400 cycles). The piezoelectric textile demonstrates multiple potential applications, including pulse wave measurement, human motion monitoring, and active voice recognition. By creatively mimicking the muscle fibers, this work paves a cost-effective way to develop high-performance and self-powered wearable bioelectronics for personalized healthcare.

1. Introduction

With the tide of 5G wireless technology, wearable biomedical electronics combined with the Internet of Things are expected to change the current reactive and disease-centric healthcare system to a personalized mode focusing on disease prevention and health promotion.¹⁻³ To fulfill this duty, developing wearable biosensors that could provide continuous physiological monitoring in a biocompatible, comfortable, and stable manner is the key challenge.⁴⁻⁵ In this respect, tremendous research efforts have been devoted to fabricating wearable biosensors in the textile forms, such as textile triboelectric nanogenerators, textile piezoelectric sensors, textile resistive sensors, textile capacitive sensors, and many others for vital signal monitoring owing to their excellent wearability and durability.⁶⁻⁸ Particularly, numerous attempts have been afforded to explore piezoelectricity-based textile wearable sensors, benefiting from their self-powered property, simple structure design, light weight, cost-effectiveness, miniaturization, and the feasibility of being blended with textiles. Piezoelectric textile sensors could convert body motions, such as pulse wave, respiration, vocal cords vibration, and limb movements, into electrical signals, providing reliable, continuous, and precise physiological information for personalized healthcare.⁹⁻¹⁵

For the piezoelectric textile fabrication, there are a wide range of inorganic and organic materials that have been demonstrated so far. For instance, inorganic ceramics, including zinc oxide,¹⁶⁻²⁷ barium titanate (BTO),¹⁸ lead zirconate titanate,¹⁹ and lead magnesium niobate-lead titanate,²⁰⁻²¹ possess a huge and high electromechanical coupling properties but suffer from various disadvantages that impede the development of wearable electronics because of their brittleness, rigidity, poor mechanical impact resistance, and toxicity.²² Meanwhile, the piezoelectric polymers, such as polyvinylidene fluoride (PVDF), polyvinylidenefluoride-trifluoroethylene or poly(vinylidenefluoride-co-hexafluoropropylene)²³⁻²⁸ hold a favorable mechanical flexibility and biocompatibility with human skin but show inferior piezoelectric properties. Given this, it is crucial to combine suitable materials for piezoelectric textile fabrications. Particularly, the incorporation of inorganic piezoelectric ceramics into an organic polymer matrix provides a creative approach to synergize the advantageous features of both organic and inorganic piezoelectric materials. However, the large mismatch of mechanical moduli between the two phases imposes a tremendous challenge to deliver the mechanical stress onto the embedded ceramic fillers. Furthermore, the
crevices and poor interfacial adhesion between the polymer matrix and the fillers also hamper the stress transfer, leading to a low electromechanical coupling efficiency.\cite{22,28,29}

Dopamine (DA), a small-molecule compound with catechol and amine group,\cite{30} has recently been considered as a robust and efficient surface building block owing to its strong interfacial adhesion strength to a variety of surfaces.\cite{31–33} The polydopamine (PDA) encapsulation around the inorganic piezoelectric ceramics improves the linkage and the anchoring strength at the interfaces between the fillers and polymer matrix by forming interfacial bonds through active surface functional groups.\cite{34,35} Enhancing the durability and stretchability under repetitive external impact and fatigue load just like the connective tissue encapsulated muscle fibers. However, in a majority of previous works,\cite{34–37} surface modulation via PDA was investigated to boost the dielectric constant and breakdown strength of polymer composites (for energy storage application) rather than the piezoelectric performance.

In contrast, our work explored the PDA modification to enhance and optimize piezoelectric properties by both experimental and theoretical investigating of the mechanism and functionalities. Inspired by the human muscle fiber, a PDA based thin, interfacial-adherent linkage was constructed between the inorganic ceramic fillers and polymer matrix, so as to improve both the mechanical strength and electromechanical coupling efficiency of the piezoelectric composite fiber. It is experimentally found that the PDA modification promotes the physical contact and stress transfer capability between the inorganic fillers and organic matrix in the composite fiber, where a 3.02 wt% PDA doping amount is able to boost the piezoelectric performance by 47%. This observation was also theoretically verified by using phase-field simulation. Toward practical application, a wearable pressure sensor was developed by the muscle inspired fiber composite with a high sensitivity of 3.95 V N$^{-1}$ and excellent durability (<3% decline after 7400 cycles). The prepared device demonstrates outstanding feasibility in active human motion monitoring, pulse detection, and voice recognition. This work not only sheds light on the fundamental understanding of the surface modification effect on the piezoelectric properties of nanocomposites, but also offers a promising route toward designing high-performance piezoelectric textile based wearable bioelectronics for personalized healthcare.

2. Results and Discussions

As schematically illustrated in Figure 1a, every movement and force load of living beings is realized by the stretch and contract of skeletal muscles consisting of highly helical architecture and extracellular matrices. It is believed that the connective tissue

![Figure 1](image_url)

**Figure 1.** Structure design and functionalities of the muscle fiber inspired piezoelectric (MFP) textile. a) Biomechanical motions enabled by skeleton muscle. b) Schematic of muscle fibers reinforced by connective tissue in human leg. c,d) Comparisons between unmodified (c) and modified (d) electrospun fibers via dopamine coating. Scale bar is 1 µm. e) Schematic illustration of as-fabricated muscle fiber inspired piezoelectric (MFP) textile. f) Photograph of MFP textile stretched by a tweezer. Scale bar is 1 cm. g) Photograph of MFP textile attached on an individual’s arm. Scale bar is 2 cm.
wrapping around the helical bundles of muscle fibers enables the external load transfer and uniform stress distribution on the muscle-connective component across several size scales (Figure 1b), which remarkably improves the mechanical strength and toughness of muscle tissues. Inspired by this unique soft biotic architecture, we designed and fabricated the electrospun nanofiber structure that is encapsulated/modified by PDA. The hydroxyl and amine of DA reacted with BTO nanoparticles through van der Waals force, and cross-linked with itself to form the PDA with high adhesion to the PVDF polymer matrix. Here, PDA is expected to mimic the function of the connective tissue around muscle fibers to enhance the robustness and durability of the as-prepared nanofiber structure. The surface modification via PDA coating affected the interfacial morphologies and properties of the electrospun piezoelectric fibrous configuration. As for the untreated BTO/PVDF nanofibers (Figure 1c), some BTO nanoparticles protrude out from the surface as nanostructured knots, which cannot efficiently involve in the electromechanical coupling process along the fibers and fails to yield piezoelectric potential inside the electrospun composite fiber. On the contrary, as illustrated in Figure 1d, the introduction of PDA coating on the surface of nanofibers wraps the protruded BTO particles and smoothens the fiber surface. Such a favorable morphology is beneficial for load transfer inside the PDA@BTO/PVDF composite fiber, leading to a stronger electromechanical coupling effect and thus a higher piezoelectric response. The electrospun nanofibers were bilaterally laminated with aluminum foils as electrodes and flexible PET substrates/superstrates to avoid unexpected impairment and contamination from the ambient environment (Figure 1e). The as-prepared device can be easily bent by a tweezer (Figure 1f), unraveling its flexible and stretchable nature for wearable sensing application. Furthermore, this muscle fiber inspired piezoelectric (MFP) textile demonstrates biocompatibility and mechanical durability that can comply with the skin’s modulus of elasticity and expediently attach to different body parts for continuous self-powered health monitoring (Figure 1g).

To evaluate the impact of PDA modification on the morphology of electrospun nanofibers, scanning electron microscope (SEM) was utilized to characterize the unmodified and modified (i.e., MFP) textiles with various BTO mass fractions. When BTO mass fraction was fixed at 3 wt%, a number of even surface was presented for the modified nanofibers with PDA coating (Figure 2b), implying that the PDA modification decreases the crevices between ceramic fillers and polymer matrix. According to Figure 2c–f, at higher BTO mass fractions, such as 5 and 7 wt%, the PDA modification efficiently eliminates the protruded knots on the surface of nanofibers. It indicates that the embedment of BTO particles inside the PDA coating is fully completed. In the as-synthesized composite nanofiber, the BTO fillers serve as an energy generation source and yield piezoelectric potential in response to external stress, whereas the PDA coating functions as a binding agent as well as stress reinforcing agent at the interface between two phases.

Figure 2g compares the X-ray diffraction (XRD) patterns of the unmodified textile and MFP textiles with different weight percents of PDA. It can be seen that the diffraction peaks at 21.9°, 31.2°, 38.5°, 44.9°, 50.4°, and 55.9° corresponds to the crystal planes of (100), (110), (111), (200), (210), and (211) for the BTO ceramic fillers in the composite film, respectively. The intensities of these peaks rise with increasing BTO mass fractions, indicating good crystallinity together with an excellent ferroelectric tetragonal phase. In addition, the typical diffraction peaks at 20.3° corresponding to the (110) and (200) crystal planes verify the existence of β-phase PVDF, while other diffraction peaks of PVDF crystal phases are not obvious, revealing the high purity of the as-synthesized MFP textile.

Figure 2h shows the Fourier-transform infrared (FTIR) spectrum of the unmodified textile and MFP textiles with various BTO mass fractions. It is noting that in comparison with the pure PVDF sample, no extra typical absorption peak in the nanofiber composite film was observed, which confirms that the addition of PDA@BTO did not change the crystal forms and molecular chain configurations of the PVDF substrate. The FTIR spectra of samples consist of tiny adsorption peaks at 763 and 976 cm⁻¹ for α-phase and large adsorption peaks at 837 and 1273 cm⁻¹ for β-phase, indicating β phase is the main crystal phase. According to the Lambert-Beer law, the F(β) values of the MFP textiles with a mass fraction of 0, 3, 5, 7, and 10 wt% are 86.8%, 73.1%, 63.0%, 63.1%, and 67.3%, respectively. The F(β) of composites is lower than that for pure PVDF. This is attributed to that the addition of BTO fillers into the PVDF polymer matrix decreases the conductivity of the precursor mixture for electrospinning and inhibits the movement of PVDF molecule chains in the process of forming β-phase, leading to the reduction of F(β).

To attain a quantitative understanding of the mechanical reinforcement and the corresponding piezoelectric performance stemming from the PDA modification, comprehensive phase-field simulation based on a Fourier spectral iterative perturbation method was performed to systematically study and compare the stress transfer capability, electromechanical coupling effect of the electrospun nanofiber architecture with and without the PDA modification. The applied external compressive stress is 1 × 10⁶ Pa, similar to our experiments. In the phase-field model, the total simulation system is discretized into a 3D array of 128 × 128 × 128 grid points with a 3D periodic boundary condition. As displayed in Figure 2i,m, two types of randomly distributed nanofibers based 3D architectures were generated by computer to mimic the MFP textile and unmodified textile with a BTO mass fraction of 5%, where the coating layer (dark) on the surface of the electrospun fibers (yellow) donates modeling structure for continuous deposition of PDA during the synthesis process.

The spatial distribution of electric field E(t), electric displacement D(r), stress σ, and strain ε in the composite under an external stress could be modeled through solving the electrostatic equilibrium equation, and the elastic equilibrium equation,

\[ \nabla \cdot D = \nabla \cdot (\varepsilon_0 \varepsilon \varepsilon_r E + d \sigma) = 0 \]  \hspace{1cm} (1)

and

\[ \nabla \cdot \sigma = \nabla \cdot (c \varepsilon - \varepsilon_0 E) = 0 \]  \hspace{1cm} (2)

where ε₀ is the vacuum dielectric permittivity, εₘ(r), d(r), and c(r) are relative dielectric permittivity, piezoelectric coefficient, and
elastic stiffness of the local phase, respectively. Equations (1) and (2) are iteratively solved using the Fourier spectral iterative perturbation method.\cite{39,40} The detailed material parameters are listed in Table S1, Supporting Information.

According to the phase-field simulation of stress distribution (Figure 2j,n) in response to a fixed vertical external stress of $1 \times 10^6$ Pa, the applied stress mainly focuses on two ends of the unmodified electrospun fibers rather than the whole surface along the fibers. This is because that the ends of the fibers possess a sufficient contact area and linkage with polymer matrix that are favorable for receiving the external force, whereas the crevices between two phases along the fibers prevent the load transfer from polymer to ceramic fillers.\cite{41} On the contrary, within the PDA coated electrospun fibrous configuration of MFP textile, the mechanical stress can effectively reach the inorganic ceramic fillers and render a more uniform and intensive stress distribution around the piezoelectric fibers (Figure 2n), just like the function of the muscle-connective
component. The larger induced stress is able to trigger a stronger electric field (Figure 2k,o) and piezoelectric potential (Figure 2l,p) based on the direct piezoelectric effect. In addition, both stress–strain test and phase-field calculation verify that the PDA coating enhanced the mechanical stiffness of MFP textile in comparison with unmodified one (Figure S1 and Table S1, Supporting Information). These results further confirm the aforementioned theoretical derivation that the DA coating is conducive to the external stress transfer onto the embedded piezoelectric ceramic and thus a stronger electromechanical coupling as well as a higher piezoelectric coefficient ($d_{33}$).

To evaluate the piezoelectric properties of the MFP textile, the output performance of the PDA modified and unmodified devices with different BTO mass fractions were systematically characterized under different stress and strain conditions, as plotted in Figure 3 and Figure S2, Supporting Information. It can be clearly seen that the MFP textile delivers higher output voltage and current than that of the unmodified devices regardless of the mass fraction of BTO in the polymer nanocomposite in response to a fixed stress of 2 N (Figure 3a,b; Figures S3 and S4, Supporting Information). It is found that both the MFP textile and unmodified textile reach maximum electric outputs under a BTO mass fraction of 5 wt% and the surface modification gives rise to a gain of $\approx 77.8\%$ and $\approx 40\%$ for output voltage and current, respectively.

Therefore, the following investigations were implemented using MFP textile with a BTO mass fraction of 5 wt%. To demonstrate the sensing performance of the MFP textile in response to the applied force, the output voltage and current were measured under various applied force from 0.07 to 2 N. As revealed in Figure 3c, the output voltage and current reveal an alternating output behavior and own a linear relationship with the applied force. A sensitivity of 3.95 V N$^{-1}$ and linearity ($R^2$) of 0.966 was achieved for MFP textile and a sensitivity of 2.26 V N$^{-1}$ and $R^2$ of 0.994 for unmodified textile (Figure 3d,e). A similar tendency was observed for the output current (Figure S5, Supporting Information). It is obvious that the MFP textiles possess higher sensitivity for both output voltage and current in comparison with that of the unmodified ones. Such a sensing behavior implies the improvement of piezoelectric performance enabled by the PDA modification. Furthermore, to verify the long-term stability and robustness of the MFP textile, cyclic loading and releasing of 5 N were performed by a linear motor at a frequency of 1 Hz. As exhibited in Figure 3f, the output voltage did not appear to decline by more than 3% appreciably after 7400 cycles, indicating an excellent mechanical durability.

To demonstrate the potential and feasibility of MFP textiles in the portable healthcare diagnosis, the fabricated device was attached to different parts of the human body for real-time physiological monitoring. Among the diversity of physiological signals, wrist pulse is considered to be one of the most common vital signs for the diagnosis of cardiovascular disease based on the evaluation of arterial blood pressure and heart rate.$^{[42]}$ The MFP textile was conformally attached above the radial artery.
near the wrist of a tester (inset of Figure 4a) to distinguish subtle variation in arterial blood pressure under different bioactive conditions. The frequency and amplitude of wrist pulses are accurately reflected in real-time with regards to the interval between two adjacent peaks and the average amplitude of peaks in the dynamic output signal profile, respectively. It is obvious that the normal resting heart rate stays around \( \approx 75 \) bpm while the dynamic heart rate after exercise reaches up to 122 bpm, as displayed in Figure 4a. The measured data are very close to the result from a commercial instrument (78 bpm) (Figure S6, Supporting Information), which verifies the capability of the prepared device in real-time physiological assessment in clinic diagnosis.

Figure 4b elucidates the enlarged view of the highlighted section of a single peak from two kinds of physiological status recorded in Figure 4a. Aside from \( P_0 \) peak as the indicator of diastolic pressure, three distinct typical peaks of the pulse waveform \((P_1-P_3)\) can be explicitly observed for both static and dynamic condition, including the peak of the early systolic pressure \((P_1(t_1))\); Percussion wave (P wave), late systolic shoulder enhancement \((P_2(t_2));\) Tidal wave (T wave) and diastolic pulse waveform \((P_3(t_3));\) Diastolic wave (D wave). As shown in Figure 4b, the average augmentation index \((A1x = P_2/P_1)\) under normal rest conditions is 0.416 \((\Delta \tau = 0.08 s)\), and the average \(A1x\) under exercise conditions is 0.459 \((\Delta \tau = 0.068 s)\). The \(A1x\) and \(\Delta \tau\) values are consistent with the value of the eye pressure. The experimental results verify that the prepared MFP textile owns an extremely fast response time to detect external stimulations and the ability to distinguish the slight variations of pulse waveform in real-time, implying the capability in monitoring the physiological conditions of the human cardiovascular system.

It is worth noting that three typical peaks of the radial artery pulse waveforms can be intuitively detected and recorded by MFP textile under the static condition and excited condition (Figure 4c), where a period of 39 and 27 ms were respectively observed for the static and excited states. Moreover, to demonstrate the capability in monitoring the slight limb movement, the MFP textile was mounted on the finger to spontaneously distinguish the bend and stretch of the detected finger. The output voltage is proportional to the bending angles from 30° to 90° (Figure 4d), revealing an outstanding durability and sensitivity to the body movement.

In regard to the cardiovascular system, the characteristics of typical feature points in pulse wave endow the detailed information for arterial elasticity, peripheral resistance, and left ventricular contractility. A set of the carotid artery pulse waveforms were detected and recorded from three candidates who worn the MFP textile on the same position of their necks (Figure 4e). Average pulse rate and distinguishable feature points \((P_1 \text{ to } P_2)\) can be explicitly obtained from the output signal profiles for each tester with diverse health status and vital signal, which reveals the wide range of applications and biocompatibility.

To verify the feasibility and possibility of MFP textile as a speech pattern recognition system, the prepared sensor was attached to the speaker’s throat so as to detect the vibration of the throat when saying different words like “Hi”, “OK”, and

![Figure 4. Device performance of the prepared MFP textile for physiological monitoring. a) Real-time output current profiles of pulse for static status and dynamic status after exercise. Inset: Photograph of the device attached to the wrist of a tester. b) Enlarged views of one cycle of electrical signal consisting of three typical peaks \((P_1, P_2, \text{ and } P_3)\) before and after exercise. c) Analysis of a single pulse wave for static status and dynamic status after exercise. d) Real-time output voltage profile of finger bending at various angles. e) Pulse waveforms of different testers when wearing the fabricated MFP textile on the same position of their necks. f) Dynamic output profile for spontaneous voice recognition when saying different words.](image-url)
“sensor”. As exhibited in Figure 4f, distinct signal profiles were acquired from the different spoken words, where the monosyllable pronunciation of “Hi” gives rise to a single peak of the recorded output voltage profile while the disyllable pronunciation of “OK” contributes to a bimodal shape. The negligible second peak of “sensor” in the output signal profile is ascribed to its weak end syllable. These results verify that the synthesized MFP textile is able to identify the tiny pronunciation variations and also reveals an excellent reproducibility, which further confirms its great potential in voice recognition in the field of biomedicine, remote-controlled human-machine interface.

3. Conclusions

In summary, we developed an MFP textile for physiological monitoring via surface modification of PDA. Combined with phase-field calculation, a comprehensive investigation was conducted to explore the mechanism and function of PDA modification in determining the piezoelectric and mechanical properties of piezoelectric composites. Both the phase-field simulation and experiment confirm that dispersing PDA into the electrospun BTO/PVDF nanofibers significantly promotes the interfacial adhesion and linkage, leading to a superior mechanical and piezoelectric performance of the as-synthesized nanocomposites. A 3.02 wt% PDA doping amount into the BTO/PVDF composites gives rise to an enhancement of 47% in piezoelectric voltage. In addition, the fabricated MFP textile is proved to be qualified for pulse monitoring, respiration identification, and voice recognition. This work not only strengthens the fundamental understanding of piezoelectric polymer composites but also offers an innovative approach in optimizing the piezoelectric energy harvesters and active sensors for wearable healthcare monitoring.

4. Experimental Section

Surface Modification Using Dopamine: 24.5 mg Tris (>99.9%(T), Aladdin, China) was dissolved into 20 mL deionized (DI) water to form Tris buffer solution (pH 8.5, 10 mM). Then, 100 mg BTO (<100 nm, 99.9%, Aladdin, China) was added and ultrasonically treated for 30 min to ensure BTO was uniformly dispersed. Subsequently, 120 mg of DA·HCl (powder, Aladdin) was added to the solution and magnetically stirred at room temperature for 24 h to form suspension. During the stirring processes, the surface of the BTO nanoparticle was coated with a PDA layer because of the self-polymerization of DA.[31] After that, the PDA@BTO nanoparticle was separated by centrifugation and then washed with DI water and anhydrous ethanol several times. Finally, the resultant was dried in an oven at 60 °C for 12 h to completely evaporate the residual solvent.

Preparation of Sensing Materials: 68 mg of the prepared PDA@BTO nanoparticles was solved into a mixture of DMF (6 mL) and acetone (4 mL). A follow-up ultrasonic treatment was conducted for 20 min to make the nanoparticles evenly dispersed in DMF solvent. Then, 2.25 g of PVDF powder was added to the mixture, followed by magnetic stirring for 2.5 h under a 50 °C water bath. This process facilitates the formation of hydrogen bond,[18] leading to a stable and homogeneous PDA@BTO/PVDF spinning precursor mixed solution.

Fabrication of Muscle Fiber Inspired Piezoelectric Textile: After the solution was stirred, 30 min ultrasonication was employed to further disperse the nanoparticles. The mixture above solution was loaded into a 10 mL BD plastic syringe that connected to a flat-tipped needle with an inner diameter of 0.5 mm. The following electrospinning conditions were utilized to obtain PDA@BTO/PVDF nanofiber film: Solution feed rate: 10 µL min⁻¹, applied voltage: 18 kV, and spinneret tip to collector distance of 15 cm. Subsequently, the flexible nanofiber composite film was cut into small pieces of 2.6 cm × 2.3 cm, and two pieces of aluminum tape (1.8 cm × 2 cm) were attached to both sides of the composite film as electrodes to form a “sandwich” configuration. Finally, the device was seal with medical tape.

Characterization and Measurement: The morphologies of samples were characterized by a field emission SEM (S-4800, Japan, HITACHI) operated at 5 kV. XRD (D8 Advance, Bruker-AXS, Germany) equipped with Cu Kα radiation (λ = 1.5418 Å and 2θ = 20°–80°) was used to analyze the crystal structure of the samples at room temperature. FTIR spectra of samples were measured by (IRAffinity-1S, SHIMADZU, Japan). A linear motor with tunable frequencies, velocities, and amplitudes was employed to bring about repeated and periodical impact on the surface of the device so as to stir up the sensor to generate a piezoelectric signal. The output voltage and current across the testing sensor were measured by Keithley 6514 electrometer system, and the data were recorded in real-time by a personal computer with Labview software. To quantitatively evaluate the condition of the cardiovascular system via the pulse waveforms measured by the MFP textile, three crucial cardiac parameters were introduced to describe the elasticity of arteries, including the time delay between the first and second peaks Δτ = t₂ − t₁, the Alx and pulse wave velocity.

Supporting Information

Supporting Information is available from the Wiley Online Library or from the author.

Acknowledgements

Y.S. acknowledges the Funds for Creative Research Groups of China (NO. 61421002), the National Natural Science Foundation of China (Grant Nos. 62074027, 61671115). J.C. acknowledges the Henry Samueili School of Engineering and Applied Science and the Department of Bioengineering at University of California, Los Angeles, for the startup support.

Conflict of Interest

The authors declare no conflict of interest.

Author Contributions

Y.S., G.X., and J.C. planned the study and guided the whole project. S.Z. and J.C. organized the data and drew the figures. J.C. submitted the manuscript and was the lead contact. Y.S., J.C., C. C., H.P., and W.L. performed the data analysis and wrote the manuscript. Y.Y., Q.G., H.T., Y.J., and Y.Z. assisted Y.S. in the thermodynamic analysis, finite element calculation, and phase-field simulation. All authors were involved in the design of this work and commented on the manuscript. Y.S., C.C., and H.P. contributed equally to the work. Informed signed consent was obtained from all the volunteers for the publication of the data and images.

Data Availability Statement

Data available on request from the authors.
Keywords
personalized healthcare, phase-field simulation, piezoelectric effect, smart textiles, wearable bioelectronics

Received: December 21, 2020
Revised: January 18, 2021
Published online:

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